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Abstract

The current study aimed using a two-experiment musculoskeletal simulation-based approach, measuring ACL biomechanics, knee joint kinematics and lower extremity joint loading to examine the effects of both a prophylactic knee sleeve on 1. a sport specific change of direction movement in female footballers and 2. a single leg landing in male footballers. Experiment 1 examined 12 female university first team level footballers (age 20.2 ± 1.34 years, height 1.61 ± 0.06 m, body mass 57.2 ± 5.6 kg) undertaking a 45° cutting movement in sleeve and no-sleeve conditions. Experiment 2 examined 10 male university first team level footballers (age 21.1 ± 1.13 years, height 1.77 ± 0.1 m, body mass 71.9 ± 8.6 kg) undertaking a single leg drop jump landing in sleeve and no-sleeve conditions. In each experiment, data was collected in a biomechanics laboratory and three-dimensional motion capture and ground reaction force information was collected. Three-dimensional kinematics, three-dimensional knee kinetics and ACL ligament forces/ strains were measured using musculoskeletal simulation, and participants were also asked to subjectively rate the knee sleeve in terms of both comfort and stability. Experiment 1 showed that the sleeve condition was associated with greater ACL strain (sleeve = 13.57% and no-sleeve = 10.26%) and forces (sleeve = 1.19BW and no-sleeve = 0.94BW). In addition, the brace condition also enhanced lateral compressive tibiofemoral (sleeve = 4.70BW and no-sleeve = 4.20BW) and total compressive tibiofemoral force (sleeve = 11.73BW and no-sleeve = 11.08BW). Finally, for the subjective ratings, participants indicated that the knee sleeve significantly improved perceived comfort and stability. Experiment 2 did not reveal and statistical differences between knee sleeve and no-sleeve conditions, nor any effects of the knee sleeve on subjective ratings of comfort or stability. Therefore, the findings from the current investigation suggest that the prophylactic knee sleeve examined in the current investigation does not appear to reduce the biomechanical parameters linked to the aetiology of knee pathologies in male/ female footballers.

Introduction

Football is regarded as the most popular sport in terms of audience and participants, with more than 200,000 professional and over 240 million amateur players globally [1]. Football like most other team sports is characterized by intermittent deceleration and landing activities requiring rapid and agile change of direction movements [2]. As both a competitive and recreational activity, football is associated with a plethora of physical benefits including enhanced cardiovascular, mental and bone health [3]. However, football is also connected with a relatively high incidence of injury [4], which has been shown to exert a significant burden on socioeconomic and healthcare systems [5]. Epidemiological investigations in professional players have shown injury rates of 8.0 per 1000 h and an average of 2.0 injuries per season [6] and 38.56 per 1000 h, at a rate of 0.85 time-loss injuries per match in recreational players [7].

One of the most commonly injured musculoskeletal structures in football is the knee [6,7], and the anterior cruciate ligament (ACL) is the most frequently injured knee ligament [8]. The ACL itself is vital for the provision of knee stability during the dynamic activities associated with football [9]. With its unique functional properties, attachment points and complex anatomy, the ACL is highly effective in restraining both excessive anterior tibial translation and coronal/ transverse plane knee motions [10]. ACL injuries in football players are predominantly, non-contact in nature, in that the ligament becomes injured without physical contact between players [11].

Physiologically, ACL injuries occur when the ligament experiences excessive tensile forces and strains [12]. As the ACL serves primarily to resist anteriorly directed tibial translation in addition to knee valgus and internal/ external rotation movements; in vivo and in vitro investigations have shown that it experiences both load and strain during activities that involve these mechanisms [13]. Aetiological investigations support this, in that the ACL is most commonly disrupted in the period immediately following foot contact, in athletic tasks involving sudden decelerations, landings and cutting manoeuvres [14]. Injury to the ACL is extremely serious in competitive players, and typically leads to long term absence from football [15]. ACL pathologies typically require reconstructive intervention using auto/allografts in order to provide sufficient stability to the injured knee to allow return to training/ competitive activities [16, 17]. Silvers & Mandelbaum [18] showed that over 250,000 ACL reconstruction interventions are undertaken each year in the US alone with average allocated costs exceeding $2 billion.

Importantly, the ACL can be associated with poor healing capacity, and the risk of a second injury is as high as 30% in the ipsilateral knee and 11% in the contralateral side [19, 20]. Even after full recovery, ACL injuries frequently lead to chronic knee pain, and athletes who experience an ACL pathology are up to ten times more susceptible to early-onset degenerative knee osteoarthritis [21], leading not only to a decline in athletic participation but also enduring disability in later life [22]. Radiographic knee osteoarthritis significantly reduces health-related quality of life, and degenerative joint disease secondary to ACL injury imposes further economic burden [23]. Similarly, it has been demonstrated that psychological as well as physical wellbeing is negatively affected, and ACL injuries have been associated with anxiety, self-esteem, pain response, depression, and feelings of decreased athletic identity [24]. Importantly, previous analyses have shown that many footballers fail to return to their previous levels of athletic function, as statistically significant performance decrements have been observed in relation to non-injured controls [25]. Concerningly, both Roos et al., [26] and Walden et al., [15] demonstrated that only 30-35% of competitive footballers remained active 3 years after suffering an ACL injury.

Because of the high incidence of ACL injuries in football players [15] and the poor-long term prognosis following injury, prophylactic interventions are therefore a key clinical priority [27]. Knee braces are external devices constructed in order to improve three-dimensional knee joint dynamic alignment [28] and range from semi-rigid devices incorporating uni or polyaxial hinges to more compliant sleeves designed simply to provide compression and enhance proprioception [29]. Knee braces represent a conservative and relatively low-cost external apparatus that are minimally invasive/ restrictive such that they can be worn during high-intensity sports maneuvers [28]. Prophylactic knee braces have been shown to reduce transverse plane knee range of motion during run, cut and vertical jump movements in netball players [28], peak knee adduction moment during a badminton lunge [30] and patellar tendon loading in run, cut and single leg hop movements in female athletes [31]. Furthermore, Sinclair et al. [32] showed using an inverse dynamics-based method of quantifying ligament loading, that ACL load rates were significantly reduced during single leg hop landings and cut movements.

However, the efficacy of any intervention modality depends on a sound comprehension of the underlying causative mechanisms of the associated condition. Inverse dynamics represent only global indices of joint loading, and therefore, are not truly representative of localized loading experienced by the joint structures [33]. Herzog et al. [34] showed that muscles are the primary contributors to the forces experienced by the lower extremity joint structures. Specifically, the complex role of muscles in controlling knee ligament loading during human movement has received insufficient attention within the literature, owing to difficulties in calculating muscle kinetics and modelling knee joint ligamentous structures [27]. To date, there has yet to be any investigation which has examined the effects of prophylactic knee bracing on ligament load and strain parameters linked to the aetiology of ACL using a muscle driven approach to quantify knee mechanics. This is principally due to the inability to non-invasively quantify ACL loads and strains during high-risk sports movements [35].

Recent, advances in musculoskeletal simulation software alongside enhancements in simulation model algorithmic complexity, mean that quantitative indices of ACL kinetics and strains are now attainable alongside more traditional simulation parameters of joint and muscle forces [36]. To date however, this more advanced modelling approach has not yet been utilized to explore the effects of prophylactic knee sleeves on ACL loading and strain during high-risk sports specific football movements. Similarly, whilst the effects of prophylactic knee sleeve have been examined previously, they have focused only on indices of knee joint loading/ kinematics. Knee sleeves are likely to mediate both kinetic and kinematic alterations at more than one body segment and thus at more than one joint; and potential positive alterations at the knee joint mediated via the sleeve, may cause concurrent effects at other lower extremity joints. Therefore, a more comprehensive approach also examining hip and ankle joint loading in addition to knee joint kinetics would be of both practical and clinical relevance.

To summarize, there is currently no scientific investigation that has explored the effects of prophylactic knee bracing on collective indices of ACL loading/ strains alongside lower extremity joint loading using musculoskeletal simulation in football players. Therefore, the aims of the current study were, using a two-experiment musculoskeletal simulation-based approach (whilst measuring ACL biomechanics, knee joint kinematics and lower extremity joint loading) to examine the effects of both a prophylactic knee sleeve on 1. a sport specific cutting movement in female university level footballers and 2. a single leg landing in male university footballers. A study of this nature may provide further insight into the comprehensive biomechanical effects of prophylactic knee sleeve designed to reduce the risk from knee pathologies in football players.

Methods

For both investigations, participants provided written informed consent and ethical approval was obtained from the University of Central Lancashire, in accordance with the principles documented in the Declaration of Helsinki. All participants were free from lower extremity musculoskeletal pathology at the time of data collection and had not undergone surgical intervention at the knee joint.

Knee sleeve

A single nylon/silicone knee sleeve (Figure 1) was utilized in this investigation, (Kuangmi 1 PC compression knee sleeve), was used in this study which came in three different sizes; small, medium and large to accommodate all participants and was worn on the dominant (right) limb in all participants. In accordance with Sinclair et al., [28], at the end of data collection participants were asked to subjectively rate the knee sleeve in relation to performing the movements without the sleeve in terms of stability and comfort. This was accomplished using 3-point scales that ranged from 1 = increased comfort, 2 = no-change and 3 = reduced comfort and 1 = increased stability, 2 = no change and 3 = increased stability.

@@@FIGURE 1 NEAR HERE@@@

Experiment 1

Participants

Twelve female (age 20.2 ± 1.34 years, height 1.61 ± 0.06 m, body mass 57.2 ± 5.6 kg and BMI = 22.1 ± 3.0 kg/m^2) university first team level footballers volunteered to take part in the current investigation.

Procedure

Participants completed five trials of a 45° cut movement in both experimental conditions (sleeve and no-sleeve). Data collection was undertaken in 22 m long biomechanics laboratory, using an a-priori approach velocity of 4.0 ± 0.2 m/s striking the force platform with their right (dominant) limb. Cut angles were measured from the centre of the force platform and the corresponding line of movement was delineated using masking tape so that it was clearly evident to participants (Figure 2). The stance phase of the cut movement was defined as the duration over > 20 N of vertical force applied to the force platform.

@@@FIGURE 2 NEAR HERE@@@

The order in which participants performed in each knee sleeve condition was counterbalanced i.e. participant 1 performed first in the knee sleeve condition followed by the no-sleeve condition whereas participant 2 was examined first in the no-sleeve condition followed by the knee sleeve and so on and so forth. To ensure consistency, each participant wore the same footwear (Asics, Patriot 6). Kinematic information was obtained using an eight-camera wall mounted motion analysis system (Qualisys Medical AB, Goteburg, Sweden) with a capture frequency of 250 Hz. The camera system was arranged in an umbrella-based configuration and covered an 8 m length and 6 m width (Figure 2). To measure ground reaction forces (GRF), an embedded piezoelectric force platform (Kistler National Instruments, Model 9281CA) operating at 1000 Hz was adopted. The GRF and kinematic information were synchronously obtained using an analogue board and interfaced using Qualisys track manager.

To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet, passive retroreflective markers of 19mm diameter were placed at the C7, T12 and xiphoid process landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS), posterior super iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal (Figure 3a). The hip, knee and ankle joint centre's were delineated according to previously established guidelines [37-39]. Carbon-fibre tracking clusters comprising of four non-linear retroreflective markers were positioned onto the thigh and shank segments. The foot segments were tracked via the calcaneus, first and fifth metatarsal, the pelvic segment using the PSIS and ASIS markers and the thorax via the T12, C7 and xiphoid markers. Static calibration trials were obtained with the participant in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters/markers, following which those not required for dynamic data were removed. The Z (transverse) axis was oriented vertically from the distal segment end to the proximal segment end. The Y (coronal) axis was oriented in the segment from posterior to anterior. Finally, the X (sagittal) axis orientation was determined using the right-hand rule and was oriented from medial to lateral (Figure 3b).

<u>@@@FIGURE 3 NEAR HERE@@@</u>

Furthermore, the effects of the prophylactic sleeve on knee joint proprioception were investigated via a weight-bearing knee joint position sense test. In accordance with the procedure of Sinclair et al. [29], (with all of the above-mentioned retroreflective markers remaining in place) participants stood in the centre of the motion capture system volume, on one leg using the dominant limb. They then slowly squatted to a knee flexion angle of 30°, which was verified using a handheld goniometer via same researcher throughout the testing process. This position was held for a period of 15 s during which time the knee 'criterion' angle was captured using the motion capture system (Figure 4ab). Following this, participants were asked to return to a standing (i.e. with both feet on the floor) position for a further 15 s, and then repeated the above process without guidance from the goniometer; a condition henceforth named 'unaided'. This position was again held for a period of 15 s and the unaided trial was similarly collected using the motion analysis system. This above process was undertaken on three occasions in both prophylactic sleeve and no-sleeve conditions using a counterbalanced order, and in between each trial participants walked a fixed distance of 20 ft to eliminate proprioceptive memory of the previous trial.

__@@@FIGURE 4 NEAR HERE@@@__

Data Processing

Dynamic and proprioception trials were digitized using Qualisys Track Manager (Qualisys Medical AB, Goteburg, Sweden) in order to identify anatomical and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). GRF data and marker trajectories were smoothed with cut-off frequencies of 50 Hz at 12 Hz respectively, using a low-pass Butterworth 4th order zero lag filter. Within Visual 3D knee joint angles were quantified using an XYZ cardan sequence (where X is the sagittal plane; Y is the coronal plane and is Z is the transverse plane).

For the proprioceptive data, the knee flexion angle during the criterion and unaided trials was calculated. The absolute difference in the knee flexion angle in degrees, was calculated between the criterion and unaided trials to provide an proprioception angular error value for both the prophylactic knee sleeve and no-sleeve conditions (with a low value indicates greater knee proprioception) and then extracted for statistical analysis. For dynamic trials obtained during the 45° cut movements, these were linearly normalized to 100 % of the stance phase. Three-dimensional angular kinematic measures from the stance phase that were extracted from the knee joint in each of the angular planes of rotation were peak angle, peak angular velocity and minimum angular velocity.

Dynamic data during the stance phase was exported from Visual 3D into OpenSim 3.3 software (Simtk.org) using a custom pipeline that allowed the inverse kinematics to be exported to match the degrees of freedom associated with the experimental model in OpenSim [27]. The standard Gait2392 Opensim musculoskeletal model was adapted to include six degrees of freedom knee joints and also an ACL bundles modelled in accordance with Sinclair et al., [27] as non-linearly elastic passive soft tissues based on the proximal (femur) and distal (tibia) insertion points of Xu et al., [40] (Figure 5ab). The model was further developed by incorporating a patella and the tibiofemoral joint was separated into medial and lateral compartment locations which were positioned at 25% and 75% of the scaled knee joint width in accordance with Barrios & Willson [41].

<u>@@@FIGURE 5 NEAR HERE@@@</u>

The model was firstly scaled within OpenSim to account for the anthropometrics of each participant, using data from the anatomical landmarks collected during the static calibration trials. In accordance with Kar & Quesada, [35], muscle and ligament dimensions were scaled in the same manner as body segments, from the static trial marker positions. Following this as muscle forces are the main determinant of joint forces [34], muscle kinetics were quantified using computed muscle control (CMC) procedure to estimate a set of muscle force patterns allowing the model to replicate the required kinematics.

Then, three-dimensional ankle, medial tibiofemoral, lateral tibiofemoral and hip joint forces as well as compressive patellofemoral joint forces were calculated via the joint reaction analyses function within OpenSim, using the muscle forces generated from the CMC process as inputs. The joint reaction analysis function in OpenSim calculates the joint loads transferred between two contacting bodies, about the joint location identified during the static trial. Furthermore, the three-dimensional forces calculated at the lateral and medial aspects of the tibiofemoral joint via the joint reaction analysis were added together in order to also determine the total tibiofemoral joint force in all three planes. In the current investigation, joint forces were normalized by dividing by each participants body weight (BW).

From the above processing, peak three-dimensional ankle, lateral tibiofemoral, medial tibiofemoral, total tibiofemoral and hip joint forces, and peak compressive patellofemoral forces during the stance phase were extracted for statistical analyses. In addition, instantaneous load rates (BW/s) for each of the aforementioned joint loads were extracted by obtaining the peak increase in force between adjacent data points and joint force impulses (BW·ms) during the stance phase were also calculated using a trapezoidal function.

In addition to the above, from the CMC process firstly the peak ACL force during the stance phase was extracted and normalized by dividing the net values by bodyweight (BW).

Furthermore, the peak forces (BW) during the stance phase for the major muscles crossing the knee joint were quantified and also the muscle force impulses (BW·ms) during the stance phase were also extracted using a trapezoidal function. In addition, the biceps femoris long head, biceps femoris short head, semitendinosus, semimembranosus muscle forces calculated via the CMC process were added together to create the total hamstring muscle force. In addition, the rectus femoris, vastus lateralis, vastus medialis and vastus intermedius forces calculated via the CMC process were also summed to create the total quadriceps muscle force. The maximum total hamstring and total quadriceps forces as well as their impulses during the stance phase were extracted for statistical analysis.

In addition, the maximum ACL strain (%) was calculated by dividing the maximum ligament bundle length during the dynamic trials by the resting length, which was obtained during the static calibration trials [35] and ACL strain rate (%/s) was by obtaining the peak increase in ACL strain between adjacent data points.

Statistical analyses

For each parameter/ condition, means and standard deviations were calculated and differences between knee sleeve and no-sleeve conditions examined using Bayesian paired t-tests with default prior scales using SPSS 27.0 software (SPSS, IBM). Bayesian factors (BF) were used to explore the extent to which the data supported the alternative (H_1) hypothesis and Bayes factors throughout were interpreted in accordance with the recommendations of Jeffreys [42] with values $\geq$3 indicating sufficient evidence in support of H_1. In the interests of conciseness and clarity only variables that presented with Bayes factors $\geq$3 are presented in the results section. Finally, using the data collected from the subjective feedback based on participants' ratings of both stability and comfort were examined using Chi-Square tests.

Experiment 2

Participants

Ten male (age 21.1 ± 1.13 years, height 1.77 ± 0.1 m, body mass 71.9 ± 8.6 kg and BMI = 22.9 ± 3.2 kg/m^2) university first team level footballers volunteered to take part in the current investigation.

Procedure

Kinematic information was obtained using the procedure and biomechanical modelling approach outlined in experiment 1 and participants once again wore the same footwear. For this experiment participants performed single leg drop jump landings with their right (dominant) limb after stepping off from a 30 cm plyometric box onto the force platform in order to simulate deceleration phase of landing [43]. The landing phase of was considered to have begun at foot contact (defined as > 20 N of vertical force applied to the force platform) and ended at the instance of maximum knee flexion.

Processing

The same processing techniques and variables as experiment 1 were adopted.

Statistical analyses

To examine biomechanical differences between conditions and subjective preferences/ ratings the same statistical analyses as experiment 1 were adopted, with the same statistical principles and reporting adhered to.

Results

Experiment 1

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Ligament biomechanics

For the peak ACL strain, values were larger in the knee sleeve (BF = 4.45) condition compared to no-sleeve (Table 1). For the peak ACL force, values were larger in the knee sleeve (BF = 25.53) condition compared to no-sleeve (Table 2).

Joint loading

For the hip shear force impulse values were larger in the knee sleeve (BF = 33.31) compared to no-sleeve (Table 1). Furthermore, for the hip medial force impulse values were larger in the knee sleeve (BF = 7.70) compared to no-sleeve (Table 1).

For the peak lateral tibiofemoral compressive force, values were larger in the knee sleeve (BF = 28.55) conditions compared to no-sleeve (Table 1). For the peak total compressive tibiofemoral force, values were greater in the knee sleeve (BF = 4.04) conditions compared to no-sleeve (Table 1).

Joint kinematics and proprioception

No differences in joint kinematics or proprioception (BF <3.0) were observed (Table 2).

Muscle forces

For peak vastus medialis force, values were larger in the knee sleeve compared to no-sleeve (BF = 3.11) (Table 3). For peak gracilis force, values were larger in the no-sleeve condition compared to the knee sleeve (BF = 5.56) (Table 3). Similarly, for the gracilis force integral, values were larger in the no-sleeve condition compared to the knee sleeve (BF = 11.81) (Table 3).

Subjective ratings

For the subjective ratings, participants indicated that the sleeve significantly improved subjective comfort ($X^2_{(2)}$ = 13.50, p<0.05) and subjective stability ($X^2_{(2)}$ = 8.33, p<0.05).

Experiment 2

@@@ TABLE 4 NEAR HERE @@@

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Ligament biomechanics

No differences in ligament biomechanics (BF <3.0) were observed (Table 4).

Joint loading

No differences in joint loading (BF <3.0) were observed (Table 4).

Joint kinematics and proprioception

No differences in joint kinematics or proprioception (BF <3.0) were observed (Table 5).

Muscle forces

No differences in muscle forces (BF <3.0) were observed (Table 6).

Subjective ratings

For the ratings of comfort, participants indicated that the sleeve did not significantly influence subjective comfort ($X^2_{(2)} = 1.75$, $p>0.05$) or stability ($X^2_{(2)} = 3.25$, $p>0.05$).

Discussion

The current investigation using a two-experiment approach, represents the first study to explore the effects of prophylactic knee bracing on ACL loading/ strains alongside lower extremity joint loading using musculoskeletal simulation in male and female football players. The debilitating nature of ACL injuries, the high rate of re-injury and the incidence of degenerative joint disease secondary to ACL injury, means that this study may provide important information necessary to inform future prevention strategies and insight into the cumulative biomechanical effects of prophylactic knee braces.

In relation to the ACL, experiment 1 showed that ACL loading and ACL strain were larger in the knee sleeve compared to no-sleeve. This observation opposes those of Sinclair et al., [31] and Sinclair et al., [32] who showed that prophylactic knee bracing attenuated knee joint soft tissue loading at the patellar tendon and ACL itself. Mechanically, aetiological analyses have shown that ACL injuries occur when the ligament itself experiences excessive tensile forces and strains [12]. Given the increases in these parameters shown in experiment 1, it appears that prophylactic knee bracing akin to that examined in this study may increase the risk from the ligamentous parameters linked to the aetiology of injury. Therefore, during the sports specific movements examined in experiments 1 and 2, the findings do not support the utilization of prophylactic knee bracing for the attenuation ACL injuries.

At the tibiofemoral joint, experiment 1 indicated that lateral and total tibiofemoral compressive loading was larger in the knee sleeve. As no-differences in medial tibiofemoral compartment loading were found it can be concluded that differences in total tibiofemoral loading were mediated through increases at the lateral tibiofemoral compartment. Whilst prophylactic knee bracing has been shown to attenuate tibiofemoral loading quantified using the peak knee adduction moment during a badminton lung30, there has yet to be an examination of the effects of knee bracing on lateral tibiofemoral kinetics. Nonetheless, despite medial tibiofemoral disorders being far more commonplace [44], the aetiology of joint degenerative pathologies is linked to excessive and habitual mechanical loading [45]. As such, experiment 1 indicates that the knee sleeve may increase the risk from the biomechanical mechanisms linked to the initiation of lateral tibiofemoral degeneration during the cut movement. Therefore, similar to the conclusions in relation to the ACL, the findings do not support the utilization of prophylactic knee bracing for the attenuation of knee joint injuries in male and female footballers during 45°cut and single leg landing conditions.

At the hip joint, the findings from experiment 1 showed that both the shear and medial force impulses were significantly larger in the knee sleeve condition compared to no-sleeve. This observation supports the principles of the walking study shown by Toriyama et al., [46], in that a knee brace significantly attenuated hip joint kinetics of the ipsilateral side. This investigation therefore highlights that knee sleeves affect joint mechanics in addition to those experienced by the knee joint itself. Thus, it is recommended that future analyses concerning knee braces, examine more than knee joint biomechanics in order to obtain a more cumulative representation of their potential prophylactic effects. Regardless, as the aetiology of hip joint degeneration is linked to the magnitude and frequency at which the applied mechanical loads are experienced [45], experiment 1 indicates that the knee sleeve may enhance the risk from the kinetic mechanisms linked to the initiation of hip joint degeneration.

Previous systematic analyses have proposed that prophylactic knee braces promote and facilitate safer landing biomechanics during functional athletic tasks by promoting an increased sensation of knee joint stability [47]. However, the subjective and proprioceptive ratings from both experiments in the current investigation provide only partial support for this notion. Experiment 1 showed that the knee sleeve enhanced subjective knee joint stability yet in experiment 2 there were no perceptual alterations as a function of the sleeve, and neither investigation showed any improvement in knee joint proprioception. It is proposed that knee braces enhance knee joint stability and proprioception by stimulating sense receptors in the skin mediated through compression provided by the brace itself [47]. However, the findings from experiment 1 do not appear to support this, as whilst improvements in perceived stability were shown, this did not translate into positive changes in knee biomechanics. It has been speculated previously that prophylactic sleeves do not provide sufficient compression to alter knee stability and proprioception sufficiently to mediate alterations in dynamic knee biomechanics [29]. Therefore, although compression provided via the knee sleeve was not examined as part of the current investigation, an interesting avenue for future analyses may be to explore devices that provide different levels of compression in regards to their prophylactic efficacy.

A potential limitation to both experiments undertaken as part of the current investigation is the mechanism by which the musculoskeletal simulation-based analyses were completed. The CMC process, although an effective and robust tool for the quantification of muscle and soft tissue kinetics utilized in previous analyses to simulate ACL mechanics [35], can be limited in its ability to quantify specific muscle coordination during dynamic tasks [48]. Furthermore, that the ACL was not modelled with sex specificity in regard to its anatomy and scaling may serve as a drawback to this investigation. Although such an approach has yet to be developed within the simulation based musculoskeletal modelling literature; as the ACL contributes pointedly to knee mechanics, incorporation of sex-specific ligament modelling may improve the efficacy of musculoskeletal simulation analyses. Finally, that only relatively modest sample sizes were utilized in both experiments may have limited statistical power and alternate statistical observations may have arisen as a function of enhanced Bayes factors with the inclusion of additional participants [49].

Conclusion

The current investigation adds to the literature by exploring via a two-experiment investigation, the effects of prophylactic knee bracing on ACL loading/ strains and lower extremity joint biomechanics using a musculoskeletal simulation-based approach in male and female footballers. This study importantly showed in experiment 1 that ACL loading/ strain, lateral and total tibiofemoral compressive forces as well as hip joint shear and medial forces were greater in the knee sleeve condition and in experiment 2 that there were no statistical effects of the knee sleeve. Therefore, the findings from the current investigation suggest that the prophylactic knee sleeve examined in the current investigation does not appear to reduce the biomechanical parameters linked to the aetiology of knee pathologies in male/ female footballers.

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Tables

Table 1: ACL and joint forces (Means ± standard deviations) for each knee sleeve condition – from experiment 1.

	Knee sleeve		No-sleeve	
	Mean	SD	**Mean**	SD
Peak ACL force (BW)	***1.19***	0.36	0.94	0.33
Peak ACL strain (%)	***13.57***	4.84	10.26	2.38
Peak ACL strain (%/s)	75.37	10.96	80.87	12.39
Peak hip compressive force (BW)	9.97	1.84	9.80	1.74
Hip compressive impulse (BW·ms)	1652.58	433.36	1708.26	452.87
Peak hip shear force (BW)	2.74	1.35	2.49	1.21
Hip shear impulse (BW·ms)	***194.20***	306.42	92.70	286.98
Hip peak medio-lateral force (BW)	***4.93***	1.15	5.85	1.10
Hip medio-lateral impulse (BW·ms)	***702.52***	301.03	830.84	310.57
Peak patellofemoral compressive force (BW)	10.08	2.45	10.17	3.00
Patellofemoral compressive impulse (BW·ms)	1350.34	465.84	1414.64	531.04
Peak medial tibiofemoral condyle compressive force (BW)	7.22	1.50	7.11	1.51
Medial tibiofemoral condyle compressive impulse (BW·ms)	1052.14	284.94	1021.79	301.41
Peak medial tibiofemoral condyle shear force (BW)	3.84	1.03	4.30	0.76
Medial tibiofemoral condyle shear impulse (BW·ms)	532.59	153.16	641.38	149.64
Peak medial tibiofemoral medio-lateral force (BW)	2.22	1.41	1.96	1.03
Peak medial tibiofemoral medio-lateral impulse (BW·ms)	306.96	196.94	265.22	195.17
Peak lateral tibiofemoral condyle compressive force (BW)	4.70	0.95	4.20	1.14
Lateral tibiofemoral condyle compressive impulse (BW·ms)	698.05	273.28	660.11	285.97
Peak lateral tibiofemoral condyle shear force (BW)	2.30	0.71	2.41	0.90
Lateral tibiofemoral condyle shear impulse (BW·ms)	316.40	149.65	334.28	163.53
Peak lateral tibiofemoral medio-lateral force (BW)	1.88	0.83	1.68	0.50
Peak lateral tibiofemoral medio-lateral impulse (BW·ms)	265.43	134.46	231.05	94.07
Peak total tibiofemoral compressive force (BW)	***11.73***	2.34	11.08	2.49

Total tibiofemoral compressive impulse (BW·ms)	1750.18	534.36	1681.90	569.46
Peak total tibiofemoral shear force (BW)	5.87	1.32	6.45	1.15
Total tibiofemoral shear impulse (BW·ms)	849.00	209.43	975.66	268.93
Peak total tibiofemoral medio-lateral force (BW)	3.79	2.27	3.31	1.48
Peak total tibiofemoral medio-lateral impulse (BW·ms)	572.39	318.22	496.27	268.97
Peak ankle compressive force (BW)	10.36	1.48	10.08	2.13
Ankle compressive impulse (BW·ms)	1525.02	387.94	1453.99	408.65
Peak ankle shear force (BW)	3.14	0.91	3.20	1.24
Ankle shear impulse (BW·ms)	191.72	237.15	100.94	255.53
Peak ankle medio-lateral force (BW)	3.96	3.96	3.94	3.94
Ankle medio-lateral impulse (BW·ms)	550.48	245.11	510.37	188.78

Notes: bold text = statistical difference between knee-sleeve and no-sleeve conditions (BF >3.00).

Table 2: Knee joint kinematics (Means ± standard deviations) for each knee brace condition – from experiment 1.

	Knee sleeve		**No-sleeve**	
	Mean	SD	**Mean**	SD
Peak knee flexion (°)	60.94	11.63	60.08	9.52
Peak knee abduction (°)	11.44	5.99	13.33	8.81
Peak knee internal rotation (°)	10.04	6.48	6.06	7.86
Peak knee flexion velocity (°/s)	505.39	70.22	464.80	113.63
Peak knee abduction velocity (°/s)	205.60	127.17	161.93	69.48
Peak knee internal rotation velocity (°/s)	288.25	150.05	308.87	108.13
Proprioception angular error (°)	3.93	1.93	4.23	1.88

Table 3: Muscle forces (Means ± standard deviations) for each knee sleeve condition – from experiment 1.

	Knee sleeve		**No-sleeve**	
	Mean	SD	**Mean**	SD
Peak biceps femoris long head force (BW)	0.49	0.31	0.46	0.33
Biceps femoris long head impulse (BW·ms)	39.60	48.66	31.17	31.03
Peak biceps femoris short-head force (BW)	0.79	0.29	0.83	0.26
Biceps femoris short head impulse (BW·ms)	60.11	36.24	59.85	28.25
Peak gracilis force (BW)	***0.14***	0.06	0.21	0.10
Gracilis impulse (BW·ms)	***7.61***	5.15	10.27	5.47
Peak lateral gastrocnemius force (BW)	1.11	0.25	1.03	0.36
Lateral gastrocnemius impulse (BW·ms)	81.85	29.55	75.65	34.28
Peak medial gastrocnemius force (BW)	2.18	0.62	2.41	0.57
Medial gastrocnemius impulse (BW·ms)	166.00	65.81	172.83	57.86
Peak rectus femoris force (BW)	2.83	0.65	2.87	0.57
Rectus femoris impulse (BW·ms)	358.71	165.51	381.55	178.20

Peak semimembranosus force (BW)	0.84	0.46	0.80	0.41
Semimembranosus impulse (BW·ms)	59.06	33.27	55.53	31.33
Peak semitendinosus force (BW)	0.27	0.10	0.27	0.11
Semitendinosus impulse (BW·ms)	15.34	7.51	15.06	7.36
Peak total hamstring force (BW)	1.80	0.73	1.61	0.61
Total hamstring impulse (BW·ms)	174.11	89.74	161.61	75.78
Peak total quadriceps force (BW)	9.80	1.92	9.39	2.21
Total quadriceps impulse (BW·ms)	1412.64	397.21	1417.13	437.73
Peak vastus intermedius force (BW)	2.61	0.48	2.46	0.70
Vastus intermedius impulse (BW·ms)	309.22	75.38	304.77	95.09
Peak vastus lateralis force (BW)	3.97	0.68	3.77	0.97
Vastus lateralis impulse (BW·ms)	457.30	121.75	450.42	149.08
Peak vastus medialis force (BW)	***2.43***	0.49	2.27	0.68
Peak vastus medialis impulse (BW·ms)	287.41	72.86	280.39	88.58

Notes: bold text = statistical difference between knee-sleeve and no-sleeve conditions (BF >3.00).

Table 4: ACL and joint forces (Means ± standard deviations) for each knee sleeve condition – from experiment 2.

	Knee sleeve		**No-sleeve**	
	Mean	SD	**Mean**	SD
Peak ACL force (BW)	0.97	0.18	0.92	0.11
Peak ACL strain (%)	12.83	3.06	11.71	1.35
Peak ACL strain (%/s)	105.94	11.57	106.67	19.57
Peak hip compressive force (BW)	9.82	2.00	10.16	1.55
Hip compressive impulse (BW·ms)	1450.31	295.85	1521.78	273.71
Peak hip shear force (BW)	2.19	0.48	2.52	0.69
Hip shear impulse (BW·ms)	302.29	118.70	368.95	153.17
Hip peak medio-lateral force (BW)	1.39	0.66	1.50	0.75
Hip medio-lateral impulse (BW·ms)	178.76	109.51	194.07	79.30
Peak patellofemoral compressive force (BW)	8.13	1.24	8.01	1.98
Patellofemoral compressive impulse (BW·ms)	1309.86	428.92	1337.48	568.47
Peak medial tibiofemoral condyle compressive force (BW)	6.83	1.61	6.80	1.04
Medial tibiofemoral condyle compressive impulse (BW·ms)	1042.57	221.16	1096.50	355.52
Peak medial tibiofemoral condyle shear force (BW)	2.69	0.26	2.70	0.52
Medial tibiofemoral condyle shear impulse (BW·ms)	424.84	131.27	409.98	140.09
Peak medial tibiofemoral medio-lateral force (BW)	0.92	0.30	0.82	0.27
Peak medial tibiofemoral medio-lateral impulse (BW·ms)	136.67	51.50	132.54	72.57
Peak lateral tibiofemoral condyle compressive force (BW)	5.22	0.95	4.65	0.56
Lateral tibiofemoral condyle compressive impulse (BW·ms)	618.42	122.87	639.73	153.60
Peak lateral tibiofemoral condyle shear force (BW)	1.84	0.38	1.82	0.46
Lateral tibiofemoral condyle shear impulse (BW·ms)	274.89	96.87	270.80	118.40
Peak lateral tibiofemoral medio-lateral force (BW)	0.32	0.15	0.30	0.08
Peak lateral tibiofemoral medio-lateral impulse (BW·ms)	27.63	19.11	25.58	17.72

Peak total tibiofemoral compressive force (BW)	11.27	1.97	10.63	0.97
Total tibiofemoral compressive impulse (BW·ms)	1660.99	312.75	1736.22	496.97
Peak total tibiofemoral shear force (BW)	4.42	0.60	4.37	0.92
Total tibiofemoral shear impulse (BW·ms)	699.73	222.50	680.78	255.84
Peak total tibiofemoral medio-lateral force (BW)	1.22	0.43	1.06	0.33
Peak total tibiofemoral medio-lateral impulse (BW·ms)	164.30	65.60	158.13	83.70
Peak ankle compressive force (BW)	8.69	1.29	8.97	1.48
Ankle compressive impulse (BW·ms)	1393.27	219.68	1442.64	333.20
Peak ankle shear force (BW)	2.33	0.57	1.99	1.29
Ankle shear impulse (BW·ms)	270.75	164.04	226.61	209.43
Peak ankle medio-lateral force (BW)	0.68	0.34	0.77	0.63
Ankle medio-lateral impulse (BW·ms)	62.85	53.41	67.67	54.34

Table 5: Knee joint kinematics (Means ± standard deviations) for each knee brace condition – from experiment 2.

	Knee sleeve		**No-sleeve**	
	Mean	SD	**Mean**	SD
Peak knee flexion (°)	65.71	7.89	66.80	8.46
Peak knee abduction (°)	4.93	3.62	3.54	3.95
Peak knee internal rotation (°)	1.66	8.46	1.78	4.35
Peak knee flexion velocity (°/s)	639.08	17.85	641.84	52.57
Peak knee abduction velocity (°/s)	102.89	41.47	159.01	50.95
Peak knee external rotation velocity (°/s)	206.35	102.86	180.05	71.63
Proprioception angular error (°)	4.13	2.39	4.42	2.15

Table 6: Muscle forces (Means ± standard deviations) for each knee sleeve condition – from experiment 2.

	Knee sleeve		**No-sleeve**	
	Mean	SD	**Mean**	SD
Peak biceps femoris long head force (BW)	0.37	0.16	0.53	0.21
Biceps femoris long head impulse (BW·ms)	39.07	33.30	44.42	19.98
Peak biceps femoris short-head force (BW)	0.37	0.19	0.55	0.27
Biceps femoris short head impulse (BW·ms)	19.88	8.22	33.72	24.87
Peak gracilis force (BW)	0.06	0.03	0.06	0.03
Gracilis impulse (BW·ms)	3.22	1.38	3.62	2.04
Peak lateral gastrocnemius force (BW)	0.50	0.16	0.73	0.31
Lateral gastrocnemius impulse (BW·ms)	44.59	16.82	62.47	32.80
Peak medial gastrocnemius force (BW)	1.20	0.34	1.69	0.66
Medial gastrocnemius impulse (BW·ms)	93.97	55.19	114.90	46.88
Peak rectus femoris force (BW)	1.96	0.33	1.90	0.36
Rectus femoris impulse (BW·ms)	161.77	40.99	176.52	36.50
Peak semimembranosus force (BW)	0.45	0.19	0.71	0.36
Semimembranosus impulse (BW·ms)	35.81	23.82	50.87	30.94

Peak semitendinosus force (BW)	0.18	0.07	0.18	0.06
Semitendinosus impulse (BW·ms)	10.04	4.61	13.89	6.55
Peak total hamstring force (BW)	1.21	0.41	1.68	0.48
Total hamstring impulse (BW·ms)	104.80	64.93	142.90	52.45
Peak total quadriceps force (BW)	7.95	1.25	7.42	1.40
Total quadriceps impulse (BW·ms)	1284.33	360.81	1285.00	532.72
Peak vastus intermedius force (BW)	2.04	0.36	1.85	0.49
Vastus intermedius impulse (BW·ms)	319.08	100.44	315.40	144.75
Peak vastus lateralis force (BW)	3.15	0.37	2.96	0.76
Vastus lateralis impulse (BW·ms)	513.36	159.53	502.71	227.52
Peak vastus medialis force (BW)	1.85	0.35	1.76	0.46
Peak vastus medialis impulse (BW·ms)	290.12	95.97	290.38	135.55

Figure labels

Figure 1: Experimental knee sleeve.

Figure 2: Experimental laboratory set-up with motion capture system cameras numbered according to the laboratory system and force platform (FP). Approach (A) and cut (C) directions are labelled with arrows showing participants direction of travel as part of the 45° cut movement.

Figure 3: a. Experimental marker locations and b. trunk, pelvis, thigh, shank and foot segments, with segment co-ordinate system axes (R = right & L = left), (TR = trunk, P = pelvis, T = thigh, S = shank & F = foot), (X = sagittal, Y = coronal & Z = transverse planes).

Figure 4: Weight-bearing knee joint position sense test from a. frontal and b. sagittal viewpoints.

Figure 5: a. Experimental Opensim model in full and b. with only the ACL bundles visible.